



Clinical assessment of chromatic aberration in phakic and pseudophakic eyes using a simple autorefractor

MARÍA S. MILLÁN,^{1,*} FIDEL VEGA,¹ FRANCISCO POYALES,² NURIA GARZÓN²

¹*Departament d'Òptica i Optometria, Universitat Politècnica de Catalunya, BARCELONATECH. Violinista Vellsolà 37, 08222 Terrassa, Spain*

²*IOA Madrid, Innova Ocular. Galileo 104, 28003 Madrid, Spain*

*m.millan@upc.edu

Abstract: We describe a psychophysical method and a simple setup – an autorefractor with a Scheiner disc, sequentially illuminated with red and blue lights – for the clinical assessment of the longitudinal chromatic aberration (LCA) in phakic and pseudophakic patients. This method applies to the unaccommodated eye, even in the presence of positive or negative refractive errors and astigmatism. It measures the chromatic difference of refraction as an estimate of LCA. We built a proof of concept from inexpensive and off-the-shelf optomechanical components with which we obtained the preliminary clinical results presented in the paper. We considered one control group of phakic patients and three groups of pseudophakic patients with monofocal implants of different designs and materials. The results, satisfactory and consistent with those reported by other researchers in related works, demonstrate the method and system feasibility.

© 2019 Optical Society of America under the terms of the [OSA Open Access Publishing Agreement](#)

1. Introduction

The optical system of the human eye is affected by a relatively high amount of longitudinal chromatic aberration (LCA) –about 2D in the visible spectral range (400-700nm) [1]. Such amount of LCA, however, has a reduced impact in vision due to several compensatory mechanisms: relatively large filtering of blue light in the lens and macular pigment and low density of the blue-sensitive cones in the cone mosaic [2–4]. The natural adaptation of human vision to LCA may explain the little interest for the clinical assessment and compensation of such aberration in phakic eyes even though binocular spatial visual acuity would benefit from bilateral correction of both spherical aberration and LCA [5,6].

LCA has recently drawn more attention with the development of advanced hybrid refractive-diffractive multifocal designs of intraocular lenses (IOLs) for refractive and cataract surgery. These designs, which combine chromatic dispersions of opposite sign between the refractive and diffractive parts of the IOL [7], may give pseudophakic patients a chance to recover their visual function with reduced chromatic aberration [8,9].

The chromatic characterization of IOLs on optical bench, using in-vitro techniques and model eyes [7,10,11] to evaluate a variety of physical metrics based on the energy efficiency and modulation transfer function (MTF), does not provide an immediate and comprehensive understanding of its impact on vision. Some researchers have developed optical setups based on either wavefront sensing, laser ray tracing, double-pass retinal images, and psychophysical techniques to precisely and effectively measure LCA *in vivo* of pseudophakic patients [12] even at various focus distances [13]. Such optical setups frequently involve the use of sophisticated equipment with strict alignment requirements, the assistance of highly specialized personnel, and the recruitment of collaborative patients for time-consuming trials. These drawbacks make it difficult to bring those techniques to the clinical practice and then, a

need arises for clinical *in vivo* assessment of LCA in pseudophakic patients. In an early contribution, Siedlecki and collaborators adapted a commercial visual refractometer to monochromatic measurements of refraction [14]. They replaced the standard white light source by a custom-designed narrow-band RGB diode, thus enabling LCA estimates through the measurement of the refractive error difference between the RGB wavelengths.

In our study, we consider a psychophysical method and a simple setup that acts as autorefractor for the clinical assessment of LCA in phakic and pseudophakic patients. We apply the Scheiner principle [15] to the observation, by an unaccommodated eye, of a distance slit illuminated sequentially with red and blue lights. A prior application of this principle to the measurement of LCA was reported by Th. Young, who modified the Porterfield's optometer also based on the Scheiner principle, and estimated LCA as early as 1801 [16,17]. In a different but related system, a Badal optometer with a Vernier target was used to precisely determine LCA and the wavelength in focus at near and far distances [18]. Other reliable methods have been reported (see, for instance, Refs. 1, 19) that do not always require from complex optical systems or strict conditions to be implemented. However, to the best of our knowledge, they have not been brought to clinics yet. To be suitable for the clinical practice, the apparatus and the examination procedure should be based on an effective *in vivo* measurement technique. Currently, these techniques apply either subjective psychophysical or objective reflectometric methods [13]. In case of considering a subjective psychophysical method, some properties are particularly valuable:

- Simple and straightforward application: based on a test easily understandable by ordinary people (the vast majority of patients are not visually trained observers);
- Inexpensive implementation, and patient-friendly driving: a few and common-use optomechanical components, easy to initialize and align, robust to the presence of common ametropia (spherical error, astigmatism), assisted by ordinary clinical personnel, and very important, quick;
- Able to provide LCA estimates in the absence of accommodation in phakic and pseudophakic patients comparable to other reported methods, rapid computation.

In our proposal—an autorefractor combined with a Scheiner disc—the chromatic refractive shift provides an estimate of LCA. This simple psychophysical method applies to an unaccommodated eye, such as in pseudophakic patients or phakic ones under mydriasis, even in the presence of positive or negative refractive errors and astigmatism. Although LCA is linearly related to the optical power of the eye, a little increase of 2.5% is predicted for each 1D of ocular power increase [19], which is commonly higher than or in the same order as the target refraction in modern cataract surgery. In general, we do not expect LCA measurements to be importantly affected by the refractive outcome of the patient. We have developed a proof of concept to obtain preliminary clinical results that may demonstrate the method and system abilities to measure LCA in the absence of accommodation. We have considered four groups of patients: one control group of phakic patients under mydriasis and three groups of pseudophakic patients with monofocal implants of different designs and materials. We report the results and discuss them in the next sections.

2. Materials, methods and patients

2.1 Optical principles for *in-vivo* assessment of LCA

The variation of the optical power of the eye with wavelength quantifies its LCA. An estimation of LCA can be obtained from the experimental measurement of the chromatic difference of refraction (CDRx) [20] from two object vergences (L_B , L_R) of the retinal conjugates of the source commonly for a short (blue, B) and a long (red, R) wavelength within the visible range

$$LCA \approx CDRx = L_B - L_R. \quad (1)$$

Although CDRx is a slight underestimation of the chromatic difference of focus in most eyes [19], many reported studies consider it as a useful estimation of LCA for practical reasons, and so we will do in this work. It is crucial to refer CDRx to the wavelength range used in the experiments before comparing results from different studies. In this study, we consider the range given by $B = 455\text{nm}$ and $R = 625\text{nm}$, which corresponds to $CDRx \approx 1.2\text{ D}$ in normal, healthy eyes, according to the results adjusted from several experimental subjective studies reported by [1].

For the measurement of CDRx in eyes with no accommodation, we combine an autorefractor with a Scheiner disc because it allows better assessment of focus than assessing blur [15]. The Scheiner disc, named after Christoph Scheiner (1619), is a double pinhole aperture with the two pinholes separated away less than the pupil diameter. The disc is placed close to the eye and centered to the pupil. The subject looks through the disc at a distance spotlight, which in our setup is an illuminated narrow slit. Figure 1 shows the basics of the Scheiner principle applied to an eye that, being emmetropic for the green light (a), becomes myopic for the blue (b) and hyperopic for the red (c) as a consequence of LCA. Thus, while the subject perceives one single green image of the slit illuminated with green light –because the two pencils converge on a common focus on the retina (a)–, they perceive two separate images of the same slit for either blue or red illumination –because the retina intercepts the two pencils after or before they converge (b, c).

Figure 2(a) depicts a scheme of the optical setup and how it would work for an ideal emmetropic eye without LCA. A white light source illuminates the slit placed at the front focal plane (F) of a lens. An achromat lens is used to not contribute to LCA. The emmetropic subject is looking through the Scheiner disc at the image of the slit formed at infinity by the lens. Such image at infinity conjugates the unaccommodated emmetropic eye's retina. Should the eye be virtually free of LCA, the two pencils would converge on a common focus on the retina and the slit would be seen singly for all wavelengths. For a normal eye affected by LCA, a refractive error raises when looking at the slit test, which appears doubled, either under blue (Fig. 1(b)) or red (Fig. 1(c)) lights.

To compensate the amount of ametropia induced by a given wavelength illumination, the subject needs to shift the slit away from F until an axial point O that conjugates – through the achromat lens and for such a wavelength- with their far point. Figures 2(b) and 2(c) show these adjustments, done for the red and the blue light, successively. In paraxial approximation, from Newton's lens formula $zz' = -f^2$, with z and z' accounting for the positions of object O and image O' from the front and back focal planes ($z = FO$, $z' = F'O'$) respectively, and f the achromat focal length, we obtain the vergence of the far point, which is given by

$$L_i = l_i^{-1} = \left[f - \left(\frac{f^2}{z_i} + k \right) \right]^{-1}, \quad i = \{R, B\}. \quad (2)$$

where k corresponds to the distance between the achromat and the cornea (Fig. 2(b)), which is assumed to be constant. We recall that distances in the direction of light propagation are considered positive, otherwise negative, in Eq. (2). If $z_i = 0$, then $L_i = 0$. From the chromatic shift (X) (Fig. 2(c)), we use Eq. (2) to calculate object vergences L_i for $i = \{R, B\}$ and substitute in Eq. (1) to obtain CDRx.

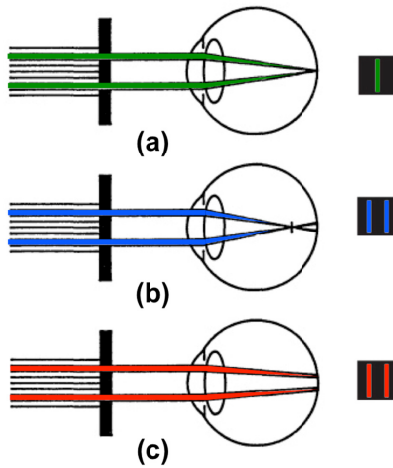


Fig. 1. Chromatic refractive error using Scheiner disc. An eye that is emmetropic for the green light (a) becomes myopic for the blue (b) and hyperopic for the red (c) as a consequence of LCA. The subject perceives a single image of a slit in (a) but two images in (b) and (c).

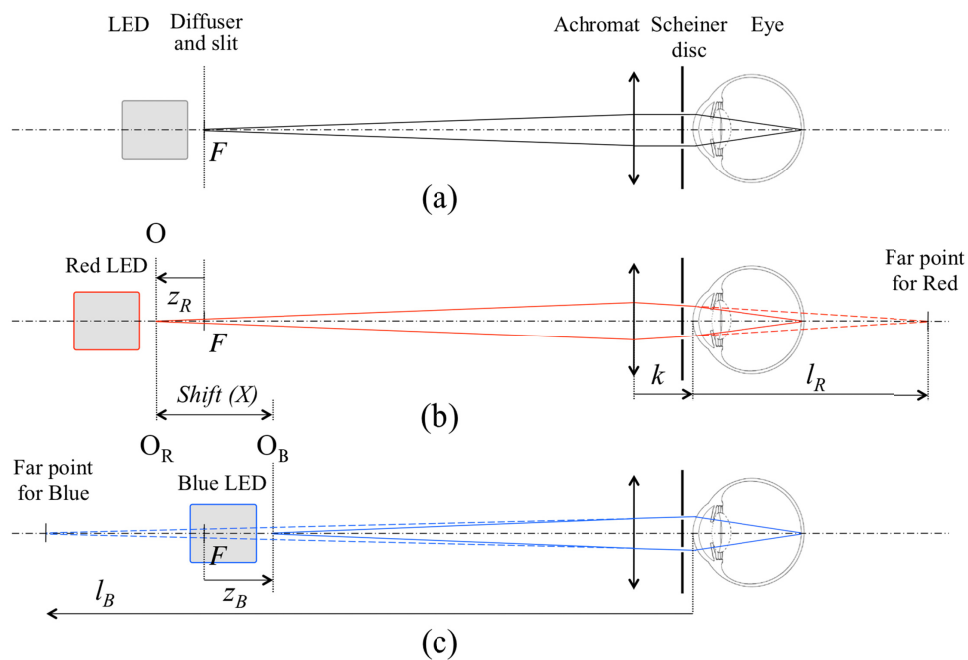


Fig. 2. Scheme of the autorefractor combined with a Scheiner disc. (a) Ideal LCA-free emmetropic subject looking at a distance test that is the image through the achromat of the slit at F . Normal emmetropic subject affected by LCA with test placed at the object conjugate through the achromat of their far point: (b) O_R , under red light, and (c) O_B , under blue light.

2.2 Experimental setup and examination procedure

Figure 3 shows a picture of the setup (proof of concept) arranged on an optical bench used in clinics. We obtained with it all the results presented in this paper. Two LEDs with emission in the blue (B) and red (R) spectral bands (Table 1) were used to alternatively illuminate the 20 μ m-width slit target. A diffuser was placed against the slit to scatter light. In the

initialization stage, we put a plane mirror behind the lens and used the autocollimation principle to precisely locate the slit at the front focal plane of the achromat lens (focal length 150mm); we adjusted the zero point of the ruler at this position (F) of the slit. A Scheiner disc consisting of two pinholes (\varnothing 0.7mm) with centers 2mm apart is close to the lens and before a head-chin rest. All the components of the setup were entirely conventional in an optical laboratory, and none of them was particularly expensive, complex or sophisticated.

Table 1. Spectral data of LEDs^a

Light	Manufacture model	λ (nm) ^a	FWHM ^b (nm)
B	Thorlabs M455L3-LED	455	18
R	Thorlabs M625L3-LED	625	18

^aNominal data from Thorlabs (see Ref. 21); ^bFull width at half maximum.

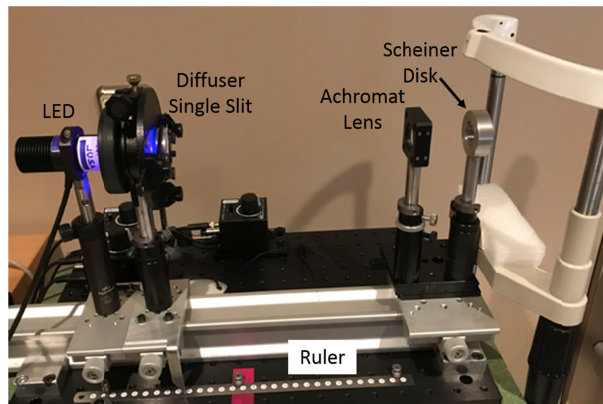


Fig. 3. Setup (proof of concept) used in clinic. Patient's head (not shown) locates at a chin rest.

An optometrist (N.G.) gave a short explanation of the test to the patient for their better understanding, collaboration and involvement. The patient was allowed to familiarize with the test by shifting the target set – consisting of the slit and LED assembled - with their own hand, forwards and backwards along the axis, while looking monocularly through the Scheiner disk with the head placed at the chin rest. Patient's eye was either monofocal pseudophakic or phakic under mydriasis to paralyze potential accommodation. The room was dimly lighted to keep the patient's pupil opened enough to allow the passage of the two pencils of light.

The measurement procedure started with the slit placed at the furthest position from the patient to make sure that they observed a double image (two lines, Fig. 1). The patient was then invited to move the target set towards them until they perceived that the two lines had just merged into a single slit image. At this moment, the optometrist read the target position on the rule (z_i) and computed the vergence for the far point position (L_i) using Eq. (2). This procedure was done with one (R or B) LED, and subsequently, repeated with the other. From Eq. (1), the difference between the vergences computed for the two R and B lights resulted in the amount of CDRx and thus in the patient's LCA. The optometrist took three measurements of CDRx from one patient's eye. The whole examination used to last about fifteen minutes per subject with the setup of Fig. 3.

2.3 Intraocular lenses

We have included three monofocal IOLs in our study: Tecnis-1 ZCB00 and Sensor-1 AAB00 (both from Johnson & Johnson Vision, Inc.) and Acrysof SN60WF (Alcon Laboratories, Inc.). The Tecnis-1 ZCB00 and the Sensor-1 AAB00 IOLs are single-piece biconvex lenses with 6.0mm aperture and overall length of 13.0mm. These IOLs are made of soft, foldable hydrophobic acrylic material with a covalently bound ultraviolet (UV) absorber. The Sensor-1

lens is spherical, whereas the Tecnis-1 model has an anterior aspheric surface that produces a spherical aberration (SA) of $-0.27\mu\text{m}$ for a 6mm pupil to compensate for the positive SA of the human cornea. The Acrysof SN60WF IOL is single-piece and composed of a copolymer of hydrophobic acrylate and methacrylate copolymer. It is a blue light-filtering biconvex sharp-edged IOL with an anterior asymmetric optic. Its aperture is 6.0mm and overall length 13.0mm. It produces a SA of $-0.20\mu\text{m}$ for 6.0mm pupil. These and further IOL parameters, such as the refractive index and Abbe number, which are relevant for the chromatic assessment of the lenses, are listed in Table 2.

Table 2. Optical data of monofocal intraocular lenses^a

	SN60WF	Tecnis-1 ZCB00	Sensar-1 AAB00
Material	Hydrophobic acrylate and methacrylate	Hydrophobic acrylic	Hydrophobic acrylic
Refractive index n	1.55	1.47	1.47
Abbe value V	37	55	55
Color filter	UV-blocking and blue light	UV-blocking	UV-blocking
Aperture (mm)	6	6	6
Design	Anterior aspheric	Anterior aspheric	Spherical
SA=c[4,0] (microns) ^b	-0.20	-0.27	NA ^c

^a Data for 555nm wavelength and 37 °C; ^b SA, spherical aberration for a 6mm pupil; ^c NA, not applicable.

2.4 Patients

CDRx was measured in 74 eyes of 74 subjects. Patients were recruited and classified into two categories: 27 were phakic and 47 were pseudophakic implanted with one of the three types of monofocal IOLs (Table 2). All patients were informed about the study and provided informed consent to undergo the clinical examinations in accordance with the tenets of the Declaration of Helsinki. The study received the approval of the local ethics committee.

We have considered two groups of phakic patients: aged >50 and aged <50 with 15 and 12 individuals respectively. Eligible patients for the study were at least 21 years of age, no cataractous eyes and no co-morbidities and regular corneal astigmatism of <0.75D. Key exclusion criteria were irregular astigmatism, ocular comorbidities, and history of ocular trauma or prior ocular surgery including refractive procedures. All experiments with phakic subjects were conducted under mydriasis to paralyze potential accommodation (tropicamide 1%, 2 drops 30 minutes before beginning the test). Table 3 shows the demographic and clinical characteristics of the phakic patients.

Table 3. Descriptive data of the phakic patients

	PHAKIC AGE>50	PHAKIC AGE<50
Number of eyes	15	12
Age (years)	61.73 ± 5.25 (54 to 72)	33.83 ± 5.27 (27 to 44)
Spherical equivalent (D)	-0.12 ± 2.03 (-3.75 to 4.50)	-0.58 ± 1.36 (-3.63 to 0.75)
Visual Acuity (Decimal) ^a	0.99 ± 0.01 (0.96 to 1.00)	0.98 ± 0.05 (0.80 to 1.00)

^a Uncorrected Distance Visual Acuity

In the case of pseudophakic patients, the inclusion criteria were: a need for cataract surgery, no ocular comorbidities and potential visual acuity after surgery higher than 0.8 Decimal. Exclusion criteria were a history of ocular disease other than cataract (e.g., uveitis, amblyopia, glaucoma), any acute or chronic condition that would increase the risk or

confound study results, any capsule or zonular abnormalities that may affect post-operative centration or tilt of the IOL and presence of pupil abnormalities.

The LCA was measured one month after surgery in 47 eyes of 47 subjects implanted with monofocal IOLs. Types of IOL implanted in the eyes of the subjects were in the following proportion: 13 eyes had the SN60WF (Alcon Inc.); 17 had the Tecnis-1 (Johnson & Johnson Vision, Inc.); 17 eyes had the Sensar-1 (Johnson & Johnson Vision, Inc.). The demographic and clinical characteristics of the pseudophakic patients enrolled in this study are listed in Table 4. There were no statistically significant differences between pseudophakic groups in terms of age, spherical equivalent, and visual acuity after surgery and IOL power.

Table 4. Descriptive data of the pseudophakic patients

	SN60WF	Tecnis-1 ZCB00	Sensar-1 AAB00
Number of eyes	13	17	17
Age (years)	71.15 ± 6.28 (64 to 87)	67.29 ± 6.37 (50 to 78)	70.82 ± 4.36 (65 to 79)
Spherical equivalent (D)	-0.29 ± 0.47 (-1.50 to 0.25)	-0.41 ± 0.85 (-2.38 to 0.75)	-0.40 ± 0.58 (-2.50 to 0.00)
Visual Acuity (Decimal) ^a	0.96 ± 0.06 (0.80 to 1.00)	1.01 ± 0.75 (0.90 to 1.20)	0.96 ± 0.07 (0.80 to 1.00)
IOL power (D)	19.46 ± 5.12 (9.00 to 27.00)	21.44 ± 2.41 (16.50 to 27.50)	20.91 ± 2.82 (17.00 to 30.00)

^aUncorrected Distance Visual Acuity

2.5 Data analysis

Data analysis was performed using the software SPSS for Windows version 22.0 (IBM, Armonk, NY, USA). Normality of data samples was evaluated by means of the Shapiro Wilk test. Average LCA and standard deviation were reported in descriptive analysis for IOLs. Independent sample *t*-tests were used to compare the differences in the LCA measurements between phakic eyes. The analysis of variables was performed using analysis of variance (ANOVA) and the differences were calculated using the multiple comparison Games-Howell test. For all statistical tests, a *p*-value of less than 0.05 was considered as statistically significance.

3. Results

The CDRx was measured for both categories of phakic eyes (two groups of patients aged >50 and <50) and pseudophakic eyes (three groups of patients implanted with either monofocal SN60WF, Tecnis-1 ZCB00, or Sensar-1 AAB00 IOL) using the optical setup and the procedure described in Section 2. The results are shown in Table 5 (phakic eyes), Table 6 (pseudophakic eyes), and Fig. 4.

Table 5. Experimental values of LCA for phakic eyes

	PHAKIC AGE>50	PHAKIC AGE<50
CDRx (D)	1.23 ± 0.42 (0.71 to 1.92)	1.35 ± 0.42 (0.69 to 2.19)

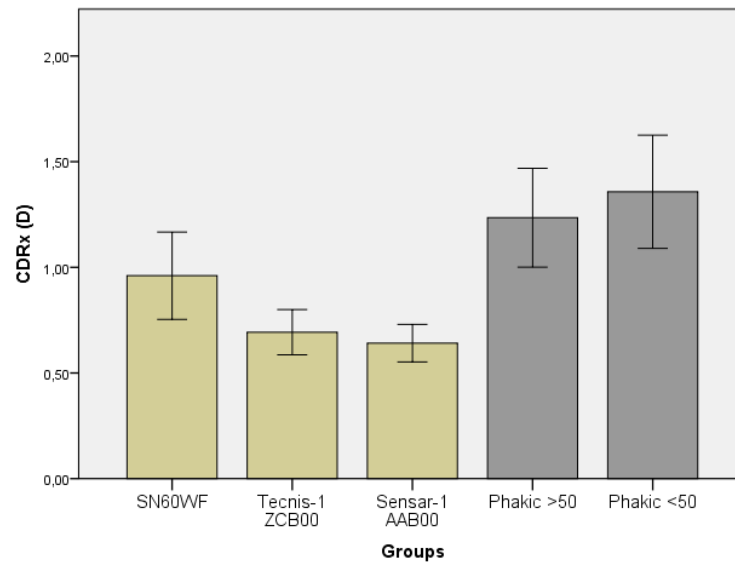


Fig. 4. Experimental CDRx obtained for the two groups of phakic eyes and three groups of pseudophakic eyes.

Table 6. Experimental values of LCA for pseudophakic eyes

	SN60WF	Tecnis-1 ZCB00	Sensar-1 AAB00
CDRx (D)	0.96 ± 0.34 (0.26 to 1.44)	0.69 ± 0.21 (0.28 to 1.04)	0.64 ± 0.17 (0.33 to 0.90)

The CDRx values obtained for the groups of phakic eyes (Table 5) were very similar ($1.23 \pm 0.42D$, for the group aged >50 and $1.35 \pm 0.42D$ for the group aged <50) and the differences between them were not found statistically significant ($p = 0.973$). The CDRx values obtained for the groups of pseudophakic eyes (Table 6) were generally lower than the values obtained for the phakics (Table 5). Such differences were found statistically significant only in the case of patients implanted with Tecnis-1 ZCB00 and Sensar-1. Comparing the CDRx values reached by pseudophakic eyes (Table 6), differences were statistically significant only between the SN60WF and Sensar-1 groups ($p = 0.032$), although the difference between SN60WF and Tecnis-1 groups ($p = 0.095$) was not far from statistical significance. Table 7 summarizes the differences in the LCA measurements between pair of groups (p-value).

Table 7. Differences in the LCA between pair of groups (p-value)

	SN60WF	TECNIS-1	SENSAR-1	Phakic < 50	Phakic > 50
SN60WF	-	0.095	0.032*	0.148	0.425
TECNIS-1	0.095	-	0.859	0.002*	0.003*
SENSAR-1	0.032*	0.859	-	0.001*	0.001*
Phakic < 50	0.148	0.002*	0.001*	-	0.973
Phakic > 50	0.425	0.003*	0.001*	0.973	-

4. Discussion

The CDRx values obtained for the groups of phakic eyes are in fairly good agreement with those of experimental studies of CDRx as a function of wavelength reported by Thibos et al. [1] and Atchison and Smith [22]. Taking into account the data for Fig. 6 in Ref. 1 and for Fig. 4 (Eq. (5a)) in Ref. 22, the CDRx corresponding to the spectral bandwidth [455nm, 625nm] is about 1.2D. In our study, we have measured $1.23 \pm 0.42D$, for the group aged >50 and $1.35 \pm 0.42D$ for the group aged <50 , with no statistically significant difference between them (Table 7). Our result is consistent with that reported by Siedlecki et al. for the control group of 10

phakic eyes ($1.12 \pm 0.14D$) in the spectral range of 470-660nm after using an adapted visual refractometer [14]. And it is also consistent with the LCA obtained by Nakajima et al. for 45 phakic eyes ($1.19 \pm 0.46D$) in a comparable spectral region of 470-660nm using a Hartmann-Shack wavefront aberrometer [23], as reported in Table 6 of Ref. 24.

Although the CDRx values we measured for the groups of phakic eyes (Table 4) were higher than those measured for the pseudophakic group implanted with SN60WF (Alcon Inc.) ($0.96 \pm 0.34D$ in Table 5), the differences were not found statistically significant. This result agrees with those reported by other researchers, such as Nakajima et al. [24], Siedlecki et al. [14], and Perez-Merino et al. [12], who conducted independent experiments to ours and used different instruments and techniques. Nakajima et al. reported the same LCA for phakic eyes and 11 eyes implanted with Alcon AcrySof series (SN60WF and SN60AT) ($1.30 \pm 0.40D$) [24]. Siedlecki et al. reported, in the same spectral range (470-660nm), CDRx of $1.17 \pm 0.52D$ and $1.12 \pm 0.14D$ ($p = 0.64$) for a group of 9 eyes implanted with SN60WF and 10 phakic eyes, respectively [14]. Perez-Merino et al. used laser ray tracing and reported LCA (Chromatic difference of focus) of $0.76 \pm 0.12D$ in the spectral region 532-785nm for a group of 9 eyes implanted with Alcon Acrysof (SN60WF) [12], which was not significantly different from the value obtained in an earlier study for a phakic population $0.78 \pm 0.16D$ using the same instrument [23].

In our study (Table 6), we obtain CDRx values for pseudophakic groups implanted with Johnson & Johnson IOLs ($0.69 \pm 0.21D$ for Tecnis ZCB00 and $0.64 \pm 0.17D$ for Sensor-1) that are lower than the CDRx values reached by the phakics ($1.23 \pm 0.42D$ for age>50-group, and $1.35 \pm 0.42D$ for age<50-group) and Alcon SN60WF pseudophakic group ($0.96 \pm 0.34D$). The differences are statistically significant between both Johnson & Johnson IOL pseudophakic groups and phakics (Table 7). The differences between the Alcon Acrysof SN60WF and the Johnson & Johnson groups are statistically significant in the case of Sensor-1 and are relatively close to significance for Tecnis-1. These results are consistent with the dispersive characteristics of the IOL materials (Abbe number of 37 for SN60WF, and 55 for Tecnis ZCB00 and Sensor-1). Our results for the Johnson & Johnson IOLs are in very good agreement with the LCA values reported in Ref. 24 for a group 16 eyes implanted with Tecnis ZCB00 group ($0.45 \pm 0.43D$) in the spectral region 470-660nm and by Pérez-Merino et al. for a group of 9 eyes implanted with Tecnis ZCB00 ($0.46 \pm 0.15D$) [12]. Pérez-Merino also reported statistically significant difference ($p<0.05$) of LCA between the Tecnis group in comparison with LCA in phakic eyes.

We obtained very similar CDRx values (magnitude and standard deviation) for the two groups of Johnson & Johnson IOLs (Tecnis-1 and Sensor-1), with no significant difference between them. This result leads us to acknowledge a negligible influence of the aspheric (Tecnis-1 ZCB00) or the spherical (Sensor-1) IOL designs in the CDRx measurements with our optical setup.

In our study, inter-subject variability and standard deviations (from 0.17D to 0.42D) are relatively large in comparison with highly precise techniques. For instance, in Ref. 26, the authors reported standard deviations of 0.04D for psychophysical techniques in the visible spectral range (488-700nm), 0.07D using retinal images, and 0.06D using wavefront sensing for the visible and near infrared spectral range (450-950nm) in the presence of natural aberrations. However, our standard deviations are comparable to those reported in [12], [14], and Table 6 of [24]. We recall that, in our case, we took six measurements per patient (three readings times two R and B wavelengths) whereas, for instance, Siedlecki and collaborators took thirty (ten readings times three R, G, and B wavelengths). Since our setup is merely a proof of concept, there is room for improvement by refining the optomechanics of the setup and the procedure, particularly, for a correct alignment of the instrument with the patient's pupil.

Our level of variability may hinder the small dependence of the chromatic difference of focus on ametropia (0.6% per diopter of axial ametropia and 2.4% per diopter of refractive ametropia, calculated for the reduced eye by Atchison et al. [27]).

Let us remark that the procedure used in this study is robust to the presence of astigmatism. The orientation of the double pinhole of the Scheiner disc selects the meridian in which the CDRx is measured. In case of astigmatism, the positions of O_R and O_B would vary with in-plane rotations of the double pinhole, but the CDRx results (Eq. (1)) would be nearly constant. The stated evidence of the negligible influence of ametropia additionally supports this fact.

Finally, in case of patients implanted with multifocal IOLs, the simultaneous perception of several double images of the slit would complicate the task and the patient would possibly require further training to accomplish the measurement of the CDRx for the different foci.

5. Conclusions

We have proved the feasibility of our method based on an autorefractor and the Scheiner disc to measure the CDRx of patients with null or relaxed accommodation in clinical practice. The results obtained for phakic and pseudophakic eyes implanted with monofocal IOLs are satisfactory and consistent with those reported by other researchers in related papers. This fact stimulates to refine the optomechanics of the setup used in this study as a proof of concept to improve its precision and patient-friendly driving. The method is robust to the presence of spherical error and astigmatism. It can be easily implemented in clinics and applied to pseudophakic patients implanted with monofocal (aspheric, spherical) IOLs. Further research is required to extend its applicability to multifocal IOL designs.

Funding

Spanish Ministerio de Economía y Competitividad y Fondos FEDER (DPI2016-76019-R).

Disclosures

The authors declare that there are no conflicts of interest related to this article.

References

1. L. N. Thibos, M. Ye, X. Zhang, and A. Bradley, "The chromatic eye: a new reduced-eye model of ocular chromatic aberration in humans," *Appl. Opt.* **31**(19), 3594–3600 (1992).
2. R. Navarro, "The optical design of the human eye: a critical review," *J. Optom.* **2**(1), 3–18 (2009).
3. R. Navarro, "Letter to the editor," *J. Optom.* **2**, 163–164 (2009).
4. P. Artal, "Optics of the eye and its impact in vision: a tutorial," *Adv. Opt. Photonics* **6**(3), 340–367 (2014).
5. P. Artal, S. Manzanera, P. Piers, and H. Weeber, "Visual effect of the combined correction of spherical and longitudinal chromatic aberrations," *Opt. Express* **18**(2), 1637–1648 (2010).
6. C. Schwarz, C. Cánovas, S. Manzanera, H. Weeber, P. M. Prieto, P. Piers, and P. Artal, "Binocular visual acuity for the correction of spherical aberration in polychromatic and monochromatic light," *J. Vis.* **14**(2):8, 1–11 (2014).
7. M. S. Millán and F. Vega, "Extended depth of focus intraocular lens: Chromatic performance," *Biomed. Opt. Express* **8**(9), 4294–4309 (2017).
8. N. López-Gil and R. Montés-Micó, "New intraocular lens for achromatizing the human eye," *J. Cataract Refract. Surg.* **33**(7), 1296–1302 (2007).
9. H. A. Weeber and P. A. Piers, "Theoretical performance of intraocular lenses correcting both spherical and chromatic aberration," *J. Refract. Surg.* **28**(1), 48–52 (2012).
10. M. S. Millán, F. Vega, and I. Ríos-López, "Polychromatic image performance of diffractive bifocal intraocular lenses: longitudinal chromatic aberration and energy efficiency," *Invest. Ophthalmol. Vis. Sci.* **57**(4), 2021–2028 (2016).
11. G. Labuz, E. Papadatou, R. Khoramnia, and G. U. Auffarth, "Longitudinal Chromatic Aberration and Polychromatic Image Quality Metrics of Intraocular Lenses," *J. Refract. Surg.* **34**(12), 832–838 (2018).
12. P. Pérez-Merino, C. Dorronsoro, L. Llorente, S. Durán, I. Jiménez-Alfaro, and S. Marcos, "In vivo chromatic aberration in eyes implanted with intraocular lenses," *Invest. Ophthalmol. Vis. Sci.* **54**(4), 2654–2661 (2013).
13. M. Vinas, A. Gonzalez-Ramos, C. Dorronsoro, V. Akondi, N. Garzon, F. Poyales, and S. Marcos, "In Vivo Measurement of Longitudinal Chromatic Aberration in Patients Implanted With Trifocal Diffractive Intraocular Lenses," *J. Refract. Surg.* **33**(11), 736–742 (2017).

14. D. Siedlecki, A. Jóźwik, M. Zając, A. Hill-Bator, and A. Turno-Kręcicka, "In vivo longitudinal chromatic aberration of pseudophakic eyes," *Optom. Vis. Sci.* **91**(2), 240–246 (2014).
15. R. B. Rabbetts, *Bennett and Rabbett's Clinical Visual Optics*, 4e (Butterworth-Heinemann Elsevier, 2007).
16. Th. Young, "On the mechanism of the eye," *Philosophical Transactions of the Royal Society of London*, 91(Part I) (1801), pp. 23–88.
17. D. A. Atchison and W. N. Charman, "Thomas Young's contribution to visual optics: The Bakerian lecture 'On the mechanism of the eye'," *J. Vis.* **10**(12):16, 1–16 (2010).
18. D. P. Cooper and P. L. Pease, "Longitudinal chromatic aberration of the human eye and wavelength in focus," *Am. J. Optom. Physiol. Opt.* **65**(2), 99–107 (1988).
19. D. A. Atchison and G. Smith, *Optics of the Human Eye* (Butterworth-Heinemann, 2000).
20. L. N. Thibos, A. Bradley, and X. X. Zhang, "Effect of ocular chromatic aberration on monocular visual performance," *Optom. Vis. Sci.* **68**(8), 599–607 (1991).
21. <https://www.thorlabs.com>.
22. D. A. Atchison and G. Smith, "Chromatic dispersions of the ocular media of human eyes," *J. Opt. Soc. Am. A* **22**(1), 29–37 (2005).
23. M. Nakajima, T. Hiraoka, Y. Hirohara, T. Oshika, and T. Mihashi, "Verification of the lack of correlation between age and longitudinal chromatic aberrations of the human eye from the visible to the infrared," *Biomed. Opt. Express* **6**(7), 2676–2694 (2015).
24. M. Nakajima, T. Hiraoka, T. Yamamoto, S. Takagi, Y. Hirohara, T. Oshika, and T. Mihashi, "Differences of Longitudinal Chromatic Aberration (LCA) between Eyes with Intraocular Lenses from Different Manufacturers," *PLoS One* **11**(6), e0156227 (2016).
25. S. Marcos, S. A. Burns, P. M. Prieto, R. Navarro, and B. Baraibar, "Investigating sources of variability of monochromatic and transverse chromatic aberrations across eyes," *Vision Res.* **41**(28), 3861–3871 (2001).
26. M. Vinas, C. Dorronsoro, D. Cortes, D. Pascual, and S. Marcos, "Longitudinal chromatic aberration of the human eye in the visible and near infrared from wavefront sensing, double-pass and psychophysics," *Biomed. Opt. Express* **6**(3), 948–962 (2015).
27. D. A. Atchison, G. Smith, and M. D. Waterworth, "Theoretical effect of refractive error and accommodation on longitudinal chromatic aberration of the human eye," *Optom. Vis. Sci.* **70**(9), 716–722 (1993).